



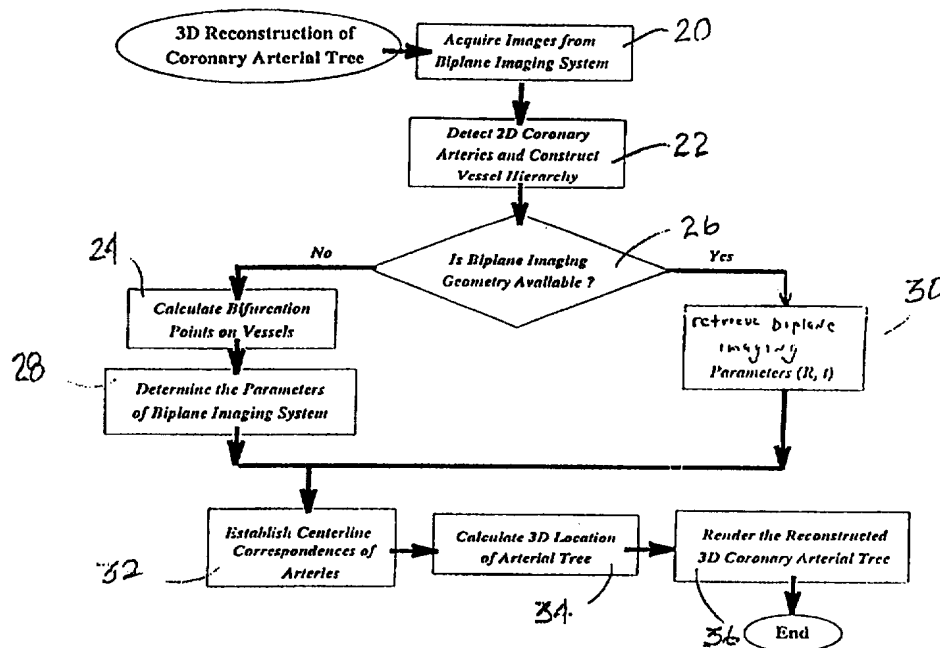
INTERNATIONAL APPLICATION PUBLISHED UNDER THE PATENT COOPERATION TREATY (PCT)

(51) International Patent Classification ⁶ : G06T 11/00		A1	(11) International Publication Number: WO 97/49065
			(43) International Publication Date: 24 December 1997 (24.12.97)
(21) International Application Number: PCT/US97/10194 (22) International Filing Date: 17 June 1997 (17.06.97) (30) Priority Data: 08/665,836 19 June 1996 (19.06.96) US (71) Applicant: ARCH DEVELOPMENT CORPORATION [US/US]; The university of Chicago, 1101 East 58th Street, Chicago, IL 60637 (US). (72) Inventors: CHEN, Shih-Yung, James; 5748 South Drexel Avenue #3B, Chicago, IL 60637 (US). CARROLL, John, D.; 4805 South Kimbark Avenue, Chicago, IL 60637 (US). METZ, Charles, E.; 165 Foxborough Place, Burr Ridge, IL 60521 (US). HOFFMANN, Kenneth, R.; 4204 Charleston Road, Matteson, IL 60442 (US). (74) Agents: SHEKLETON, Gerald, T. et al.; Welsh & Katz, Ltd., 22nd floor, 120 South Riverside Plaza, Chicago, IL 60606- 3907 (US).		(81) Designated States: AL, AU, BA, BB, BG, BR, CA, CN, CZ, EE, FI, GE, GH, HU, IL, IS, JP, KE, KP, KR, KZ, LC, LK, LR, LS, LT, LV, MG, MK, MN, MW, MX, NO, NZ, PL, PT, RO, SD, SG, SI, SK, SL, TR, TT, UA, UG, UZ, VN, YU, ZW, ARIPO patent (GH, KE, LS, MW, SD, SZ, UG, ZW), Eurasian patent (AM, AZ, BY, KG, KZ, MD, RU, TJ, TM), European patent (AT, BE, CH, DE, DK, ES, FI, FR, GB, GR, IE, IT, LU, MC, NL, PT, SE), OAPI patent (BF, BJ, CF, CG, CI, CM, GA, GN, ML, MR, NE, SN, TD, TG). Published <i>With international search report.</i> <i>Before the expiration of the time limit for amending the claims and to be republished in the event of the receipt of amendments.</i>	

(54) Title: METHOD AND APPARATUS FOR THREE-DIMENSIONAL RECONSTRUCTION OF CORONARY VESSELS FROM ANGIOGRAPHIC IMAGES

(57) Abstract

A method for in-room computer reconstruction of a three-dimensional (3-D) coronary arterial tree from routine biplane angiograms acquired at arbitrary angles and without using calibration objects. The method includes eight major steps: (1) acquiring biplane projection images of the coronary structure, (2) detecting, segmenting and identifying vessel centerlines and constructing a vessel hierarchy representation, (3) calculating bifurcation points and measuring vessel diameters in coronary angiograms if biplane imaging geometry data is not available, (4) determining biplane imaging parameters in terms of a rotation matrix R and a unit translation vector \vec{t} based on the identified bifurcation points, (5) retrieving imaging parameters if biplane imaging geometry data is already known, (6) establishing the centerline correspondences of the two-dimensional arterial representations, (7) calculating and recovering the 3-D coronary arterial tree based on the calculated biplane imaging parameters, correspondences of vessel centerlines, and vessel diameters, and (8) rendering the reconstructed 3-D coronary tree and estimating an optimal view of the vasculature to minimize vessel overlap and vessel foreshortening.



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1 **METHOD AND APPARATUS FOR THREE-DIMENSIONAL RECONSTRUCTION**
2 **OF CORONARY VESSELS FROM ANGIOGRAPHIC IMAGES**

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7 rights whatsoever.

8 **BACKGROUND OF THE INVENTION**

9 The present invention relates generally to a method for reconstructing images of
10 coronary vessels and more specifically to a method for three-dimensional (3-D)
11 reconstruction of coronary vessels from two two-dimensional biplane projection images.

12 Several investigators have reported computer assisted methods for estimation of the
13 3-D coronary arteries from biplane projection data. These known methods are based on the
14 *known* or *standard* X-ray geometry of the projections, placement of landmarks, *known* vessel
15 shape, and on iterative identification of matching structures in two or more views. Such
16 methods are described in a publication entitled "3-D digital subtraction angiography," *IEEE*
17 *Trans. Med. Imag.*, vol. MI-1, pp. 152-158, 1982 by H.C. Kim, B.G. Min, T.S. Lee, *et*.
18 *al.* and in a publication entitled "Methods for evaluating cardiac wall motion in 3-D using
19 bifurcation points of the coronary arterial tree," *Invest. Radiology*, Jan.-Feb. pp. 47-56, 1983
20 by M.J. Potel, J.M. Rubin, and S. A. Mackay, *et al.* Because the computation was designed

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1 for predefined views only, it is not suitable to solve the reconstruction problem on the basis
2 of two projection images acquired at *arbitrary* and *unknown* relative orientations.

3 Another known method is based on motion and multiple views acquired in a
4 single-plane imaging system. Such a method is described in a publication entitled
5 "Reconstructing the 3-d medial axes of coronary arteries in single-view cineangiograms,"
6 *IEEE Trans. MI*, vol. 13, no. 1, pp. 48-60, 1994 by T.V. Nguyen and J. Sklansky uses
7 motion transformations of the heart model. However, the motion transformations of the
8 heart model consist only of rotation and scaling. By incorporation of the center-referenced
9 method, initial depth coordinates, and center coordinates, a 3-D skeleton of the coronary
10 arteries was obtained. However, the real heart motion during the contraction involves five
11 specific movements: translation, rotation, wringing, accordion-like motion, and movement
12 toward the center of the ventricular chamber. Therefore, the model employed is not general
13 enough to portray the true motion of the heart, especially toward the end-systole.

14 Knowledge-based or rule-based systems have been proposed for 3-D reconstruction
15 of coronary arteries by use of a vascular network model. One such knowledge-based system
16 is described in a publication entitled "An expert system for the labeling and 3-D
17 reconstruction of the coronary arteries from two projections," *International Journal of*
18 *Imaging*, Vol. 5, No. 2-3, pp. 145-154, 1990 by Smets, Vandewerf, Suctens, and
19 Oosterlinck. Because the rules or knowledge base were organized for certain specific
20 conditions, it does not generalize the 3-D reconstruction process to arbitrary projection data.
21 In other knowledge-based systems, the 3-D coronary arteries were reconstructed from a set
22 of X-ray perspective projections by use of an algorithm from computed tomography. Due
23 to the motion of the heart and only a limited number of projections (four or six), accurate
24 reconstruction and quantitative measurement are not easily achieved.

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1 Angiograms of fifteen patients were analyzed in which two cases are selected for
2 discussion hereinafter. The biplane imaging geometry was first determined without a
3 calibration object, and the 3-D coronary arterial trees were reconstructed, including both left
4 and right coronary artery systems. Various two-dimensional (2-D) projection images of the
5 reconstructed 3-D coronary arterial tree were generated and compared to other viewing
6 angles obtained in the actual patient study. Similarity between the real and reconstructed
7 arterial structures was excellent. The accuracy of this method was evaluated by using a
8 computer-simulated coronary arterial tree. Root-mean-square (RMS) errors in the 3-D
9 position and the 3-D configuration of vessel centerlines and in the angles defining the R
10 matrix and \bar{t} vector were 0.9 - 5.5 mm, 0.7 - 1.0 mm, and less than 1.5 and 2.0 degrees,
11 respectively, when using 2-D vessel centerlines with RMS normally distributed errors varying
12 from 0.4 - 4.2 pixels (0.25 - 1.26 mm).

13 More specifically, the method for three-dimensional reconstruction of a target object
14 from two-dimensional images involves a target object having a plurality of branch-like
15 vessels. The method includes the steps of: a) placing the target object in a position to be
16 scanned by an imaging system, the imaging system having a plurality of imaging portions;
17 b) acquiring a plurality of projection images of the target object, each imaging portion
18 providing a projection image of the target object, each imaging portion disposed at an
19 unknown orientation relative to the other imaging portions; c) identifying two-dimensional
20 vessel centerlines for a predetermined number of the vessels in each of the projection images;
21 d) creating a vessel hierarchy data structure for each projection image, the data structure
22 including the identified two-dimensional vessel centerlines defined by a plurality of data
23 points in the vessel hierarchy data structure; e) calculating a predetermined number of
24 bifurcation points for each projection image by traversing the corresponding vessel hierarchy

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1 data structure, the bifurcation points defined by intersections of the two-dimensional vessel
2 centerlines; f) determining a transformation in the form of a rotation matrix and a translation
3 vector utilizing the bifurcation points corresponding to each of the projections images, the
4 rotation matrix, and the translation vector representing imaging parameters corresponding to
5 the relative orientations of the imaging portions of the imaging system; g) utilizing the data
6 points and the transformation to establish a correspondence between the two-dimensional
7 vessel centerlines corresponding to each of the projection images such that each data point
8 corresponding to one projection image is linked to a data point corresponding to the other
9 projection images, the linked data points representing an identical location in the vessel of
10 the target object after the projections; h) calculating three-dimensional vessel centerlines
11 utilizing the two-dimensional vessel centerlines and the correspondence between the data
12 points of the two-dimensional vessel centerlines; and i) reconstructing a three-dimensional
13 visual representation of the target object based on the three-dimensional vessel centerlines and
14 diameters of each vessel estimated along the three-dimensional centerline of each vessel; and
15 j) determining the optimal view of the vessel segments with minimal vessel foreshortening.

16 **BRIEF DESCRIPTION OF THE DRAWINGS**

17 The features of the present invention which are believed to be novel are set forth with
18 particularity in the appended claims. The invention, together with further objects and
19 advantages thereof, may best be understood by reference to the following description in
20 conjunction with the accompanying drawings.

21 Fig. 1 is a high level flowchart illustrating the steps according to a specific
22 embodiment of the present inventive method;

23 Figs. 2A and 2B are schematic views of the imaging system particularly showing the
24 position of the isocenter;

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1 Figs. 3A-3B are orthogonal views of typical biplane projection images based on a
2 computer simulated coronary arterial tree where each vessel centerline is represented by a
3 sequence of data points;

4 Fig. 4 is a schematic view of a specific embodiment of a biplane imaging system
5 model for defining a 3-D object point in the 3-D coordinate systems xyz and $x'y'z'$ and their
6 projections in the 2-D image plane coordinate systems uv and $u'v'$, respectively, according
7 to the present invention;

8 Fig. 5 is a schematic view showing two initial solutions used for searching the optimal
9 solutions of imaging parameters as well as the 3-D object in the given biplane systems;

10 Fig. 6 is a schematic view showing cone-shape bounding regions associated with the
11 calculated 3-D points; and

12 Fig. 7 is a schematic view showing two set-ups of a biplane imaging system resulting
13 from the employed two initial conditions yielding two sets of reconstructed 3-D objects A' -
14 D' and $A - D$ (real 3-D object points) as shown in gray and black circles, respectively.

15 DETAILED DESCRIPTION OF THE INVENTION

16 Referring now to Fig. 1, the present invention method includes eight major steps: (1)
17 acquiring biplane projection images of the coronary structure, as shown in step 20, (2)
18 detecting, segmenting, and identifying vessel centerlines and constructing a vessel hierarchy
19 representation, as illustrated in step 22, (3) calculating bifurcation points and measuring
20 vessel diameters in coronary angiograms, as shown in step 24, but only if biplane imaging
21 geometry data is not available, as shown by the "no" branch of step 26, (4) determining
22 biplane imaging parameters in terms of a rotation matrix R and a unit translation vector \bar{t}
23 based on the identified bifurcation points, as illustrated in step 28, (5) retrieving imaging
24 parameters, as shown in step 30, but only if known biplane imaging geometry data is

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1 available, as shown by the "yes" branch of step 26, (6) establishing the centerline
2 correspondences of the two-dimensional arterial representations, as shown in step 32, (7)
3 calculating and recovering the 3-D spacial information of the coronary arterial tree based on
4 the calculated biplane imaging parameters, the correspondences of vessel centerlines, and the
5 vessel diameters, as illustrated in step 34, and (8) rendering the reconstructed 3-D coronary
6 tree and estimating an optimal view of the vasculature to minimize vessel overlap and vessel
7 foreshortening, as shown in step 36. The above-described steps will be described in greater
8 detail herewith.

9 Acquiring Images From The Biplane Imaging System

10 As shown in step 20 of Fig. 1, biplane projection images are acquired using an X-ray
11 based imaging system. However, other non-X-ray based imaging systems may be used, as
12 will be described hereinafter. Such X-ray based images are preferably created using a
13 biplane imaging system where two projection images are produced substantially
14 simultaneously. The patient is placed in a position so that the target object, in this illustrated
15 embodiment, the heart, is scanned by the imaging system. The imaging system preferably
16 includes a plurality of imaging portions where each imaging portion provides a projection
17 image of the coronary structure. Due to interference between the imaging portions during
18 X-ray emission, one image portion must be turned off while the other image portion is
19 active. However, the duration of exposure is extremely short so that the two images are
20 sequentially taken such that the heart does not significantly move during the imaging period.
21 Such an imaging system may be, for example, a Seimens BICORE system. It is the time
22 between exposures that must be short. Otherwise blurring of the vessels may result. The
23 motion of the heart is significant throughout most of the heart cycle, thus most investigators
24 use end-diastole.

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1 Referring now to Fig. 1, and Figs. 2A-2B, Figs. 2A-2B schematically illustrate a
2 typical imaging system configuration where only one gantry arm is shown for clarity. An
3 X-ray source is located at the focus of a cone shaped X-ray beam which diverges outwardly
4 toward an X-ray image intensifier (image plane or view). The X-ray beam passes through
5 the target object and is received by the X-ray image intensifier. In a biplane system, two
6 such X-ray sources and image intensifiers are present. The gantry arm is rotatably mounted
7 such that the X-ray source and the image intensifier move relative to the target object but
8 always remain fixed relative to each other. The gantry arm permits the X-ray source and the
9 image intensifier to be positioned about the target object in a predetermined orientation.

10 The present invention is not limited to a biplane imaging system having only two
11 imaging portions. A multi-plane imaging system may be used having a plurality of imaging
12 portions such as, for example, three, four, or more imaging portions, each providing a
13 projection image of the coronary structure. Such an imaging system is essentially limited by
14 the size of the system relative to the treatment facility. The scope of the present inventive
15 method includes use of a system providing two or more projection images.

16 The projection images may or may not include orientation information describing the
17 relative geometry between the gantry arms. Even if the individual gantry position
18 information is available, the derivation of relative orientation based on the known gantry
19 positions becomes a non-trivial process especially when the two isocenters are not aligned.
20 A significant feature of the present inventive method permits reconstruction of a 3-D model
21 of the target object even when the relative orientation of the gantry arms is unknown. Other
22 prior art methods for reconstruction the 3-D model requires either known relative orientation
23 between the two views, known individual gantry position (resulting in ten parameters for

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1 biplane imaging geometry which may be sufficiently defined by five parameters), or at least
2 more that eight pairs of accurate input corresponding points.

3 The projection images are digitized and are entered into a work station. The
4 projection images are represented by a series of individual pixels limited only by the
5 resolution of the imaging system and the memory of the work station. The work station, for
6 example, may be an IBM RISC/6000 work station or a Silicon Graphics Indigo-2/High
7 Impact work station or the like. An input device, such as a keyboard, a mouse, a light pen,
8 or other known input devices is included to permit the operator to enter data.

9 Segmentation and Feature Extraction of the Two-Dimensional Coronary Arterial Tree

10 False detection of arteries is inevitable using a fully automatic vessel tracker,
11 especially when vessels overlap or the signal-to-noise ratio of the angiogram is low. In the
12 present inventive method, a semi-automatic system based on a technique using a deformation
13 model is employed for the identification of the 2-D coronary arterial tree in the angiograms,
14 as will be described in greater detail hereinafter. The required user interaction involves only
15 the indication of several points inside the lumen, near the projection of vessel centerline in
16 the angiogram, as is illustrated in step 22 of Fig. 1.

17 Referring now to Figs. 3A and 3B, computer simulated biplane projection images are
18 shown. Typically, the operator or the physician inspects the digitized biplane projection
19 images. Using a mouse or other data entry device, the physician marks a series of points
20 (data points) within a major vessel to define the initial two-dimensional centerline of the
21 vessel. After the major vessel has been marked, five additional branching vessels are marked
22 in the same manner such that a total of six two-dimensional centerlines are identified. Once
23 the major vessel has been marked, the remaining five vessels may be identified and marked

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1 in any order. Note, for purposes of clarity only, Figs. 3B and 3B show one major vessel and
2 only four branching vessels.

3 The above process is then performed for the other biplane projection image(s).
4 Again, the operator or the physician identifies and marks the major vessel and then identifies
5 and marks the five additional branching vessels. The branching vessels may be identified and
6 marked in any order, which may be different from the order marked in the first biplane
7 projection image. The physician must mark the same six vessels in each biplane projection
8 image. The present inventive method renders excellent results with use of only six identified
9 vessel centerlines. Other known systems require many more identified data points. An
10 example of such a system is described in a publication entitled "Assessment of Diffuse
11 Coronary Artery Disease by Quantitative Analysis of Coronary Morphology Based Upon 3-D
12 Reconstruction from Biplane Angiograms," IEEE Trans. on Med. Imag., Vol. 14, No. 2,
13 pp. 230-241, 1995 by A. Wahle and E. Wellnhofer et. al. This system requires at least ten
14 or more identified corresponding points due to ten variables that need to be optimized for
15 determination of biplane imaging parameters. Such a burdensome requirement significantly
16 increases computer processing time. Since the derived objective function employs five
17 variables to characterize the biplane imaging geometry, it only requires five corresponding
18 points.

19 Next, by use of the deformation model and ridge-point operator, described
20 hereinafter, the initially identified centerline is gradually deformed and made to finally reside
21 on the actual centerline of vessel.

Deformation model

The behavior of the deformation contour is controlled by internal and external forces. The internal forces serve as a smoothness constraint and the external forces guide the active contour toward image features. A deformable contour can be defined as a mapping of a material coordinate $s \in [0, 1]$ into R^2 .

$$\mathbf{v}(s) = (x(s), y(s))$$

Its associated energy function can be defined as the sum of an internal energy and an external energy. The external energy accounts for the image force F_{img} , such as a line or edge extracted from the image content, and other external constraint forces F_{ext} , such as a pulling force intentionally added by the user. The energy function is the sum of the internal energy and the external energy and is written as:

$$\begin{aligned} E(\mathbf{v}) &= E_{int}(\mathbf{v}) + E_{ext}(\mathbf{v}) \\ &= E_{int}(\mathbf{v}) + E_{img}(\mathbf{v}) + E_{ext}(\mathbf{v}) \end{aligned}$$

Equ. (A)

The internal energy, which is caused by stretching and bending, characterizes the deformable material and is modeled as:

$$E_{int}(\mathbf{v}) = \frac{1}{2} \int_0^1 \{ \alpha(s) |\mathbf{v}'(s)|^2 + \beta(s) |\mathbf{v}''(s)|^2 \} ds$$

Equ. (B)

where $\alpha(s)$ and $\beta(s)$ control the tension and the rigidity at point $\mathbf{v}(s)$. The first order term, measuring the length of ds , resists stretching, while the second order term, measuring the curvature at ds , resists bending.

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Let $\Delta I(v)$ denote the image force at point $v(s)$, which is the directional maximum (or minimum) response of gray level in a region with m by m pixels. These point sources of force are referred to as *ridge* points which will act on the contour. Based on the image force, the image energy is defined as

$$E_{img}(\mathbf{v}) = -\frac{1}{2} \int_0^1 |\Delta I(\mathbf{v}(s))|^2 ds$$

Equ. (C)

Here, only the ridge points are considered as the external force. The shape of contour under the forces becomes a problem of minimization of the energy function.

$$E(\mathbf{v}) = \frac{1}{2} \int_0^1 \{ \alpha(s) |\mathbf{v}'(s)|^2 + \beta(s) |\mathbf{v}''(s)|^2 - |\Delta I(\mathbf{v}(s))|^2 \} ds$$

Equ. (D)

A necessary condition for a contour function to minimize Equ. (D) is that it must satisfy the following Euler-Lagrange equation:

$$-(\alpha \mathbf{v}')' + (\beta \mathbf{v}'')'' + F_{img}(\mathbf{v}) = 0$$

Equ. (E)

When all the forces that act on the contour are balanced, the shape change on the contour is negligible and results in an equilibrium state.

The deformation process starts from an initial contour. The contour modifies its shape dynamically according to the force fields described above until it reaches an equilibrium state. The user manually indicates points near the vessel centerline and a spline-curve is formed based on the selected points. This serves as the initial centerline of

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1 the vessel. Without loss of generality, the artery is imaged as dark intensity against the
2 background in the angiogram. According to the densitometric profile model, the 2-D
3 cross-sectional profile of the coronary arteries has a minimum intensity at the vessel
4 centerline. An m by m operator is convolved with a given arterial image by which the ridge
5 pixel is identified if it is a directional minimum on intensity. By use of the deformation
6 model, the set of ridge pixels serves as the external forces which act on the initial model
7 curve such that it will gradually be deformed and finally reside on the real centerline of the
8 vessel.

9 A computer-based editing tool is employed for the correction in case a false-negative
10 or false-positive detection occurs. The editing tool provides the operator with the ability to
11 change the shape of the vessel centerline. This is done by the modification of control points,
12 such as addition, deletion, or dragging of the points on a spline-based curve that models the
13 vessel centerline.

14 The identified centerlines and the branching relationships are used to construct the
15 hierarchy in each image by their labeling according to the appropriate anatomy of the
16 primary and secondary coronary arteries. The labeling process on the coronary tree is
17 performed automatically by application of the *breadth-first search* algorithm to traverse
18 identified vessel centerlines, as is known in the art. From each vessel of the coronary tree
19 that is currently visited, this approach searches as broadly as possible by next visiting all of
20 the vessel centerlines that are adjacent to it. This finally results in a vessel hierarchically
21 directed graph (digraph) containing a set of nodes corresponding to each individual artery and
22 two types of arcs (descendant and sibling arcs) defining the coronary anatomy.

23 In addition to the coordinates of the vessel centerlines, the relative diameters of the
24 vessels are determined. The diameter of each vessel is estimated based on the maximum

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1 vessel diameter at a beginning portion of the vessel and a minimum vessel diameter at an
2 ending portion of the vessel. This step is performed within step 22 (Fig. 1) and is also
3 performed by the operator or the physician. The physician measures the minimum and
4 maximum vessel diameter on the projection image at the beginning of a vessel and at the end
5 of the vessel and enters the data into the work station. Only the minimum and maximum
6 diameters are required since typical values of vessel taper are known and are substantially
7 constant from patient to patient. For example, a vessel may have a maximum diameter of
8 0.35 mm at the proximal RCA and a minimum diameter of 0.02 mm at the distal RCA. The
9 remaining diameters between the two points can be calculated by linear interpolation based
10 on the maximum and minimum diameters.

11 Next, a determination is made whether biplane imaging geometry is available, as
12 illustrated in step 26 of Fig. 1. As described above, the geometric orientation of the gantries
13 during exposure may not be available or alternately, if it is available, may require a
14 calibration process. Based on the current imaging technology, the information of a single
15 plane system includes the gantry orientation (LAO and CAUD angles), SID (focal spot to
16 image intensified distance), and magnification. However, such information is defined based
17 on each individual reference system. In other words, the relative orientation that
18 characterizes the two views is unknown. Therefore, it is necessary to determine the biplane
19 geometry. If the two reference points, which are the location of the iso-centers, are made
20 to coincide, the relative orientation can be calculated directly from the recorded information.
21 However, such coincidence of the reference points is difficult to achieve in a practical
22 environment. If the accurate relative orientation data is available from the mechanical
23 hardware of the gantry, steps 32-36 of Fig. 1 may be employed to calculate the 3-D coronary
24 arterial structures. However, it is difficult to obtain biplane transformation data based on

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1 current instrumental technology. A significant advantage of the present inventive method is
2 that 3-D reconstruction is accurately rendered from 2-D projection images when such
3 orientation information is unavailable.

4 If biplane imaging geometry is not available, the bifurcation points are calculated, as
5 shown in step 24 of Fig. 1. An important step in the present inventive method relies on the
6 accurate establishment of correspondence between image features, such as points or curve
7 segments between projections. The bifurcation points on the vascular tree are prominent
8 features and are often recognized in both images to facilitate the determination of biplane
9 imaging geometry. Using the hierarchical digraph, bifurcation points are then calculated by
10 use of each pair of ascendant and descendant nodes (or vessels). Given two sets of 2-D
11 points representing the respective centerlines of vessels constituting a bifurcation, they can
12 be modeled as two curves, $p(r)$ and $q(t)$, where $0 \leq r, t \leq 1$ are the parameters based on
13 the spline-based curve-fitting algorithm, such as is described by R.H. Bartels, J.C. Beatty,
14 and B.A. Barsky in a publication entitled "*An introduction to splines for use in computer*
15 *graphics and geometric modeling*," Morgan Kaufmann Publishers Inc., Los Altos, California,
16 as is known in the art. Let curve $q(t)$ denote the branch of the primary vessel as modeled
17 by a curve, $p(r)$. The bifurcation point can then be obtained by calculation of the intersection
18 of the tangent line denoted as a vector q_0 at the point $q(0)$ and the curve of the primary
19 vessel, $p(r)$, by minimizing the objective function $\mathcal{F}_{(bif)}(r)$ as follows:

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$$\min_r \mathcal{J}_{\{bif\}}(r) = \left\| \mathbf{p}(r) - \frac{\mathbf{q}_O^T \mathbf{p}(r)}{\mathbf{q}_O^T \mathbf{q}_O} \mathbf{q}_O \right\|^2$$

$$= \frac{(\mathbf{p}^T(r) \mathbf{p}(r)) (\mathbf{q}_O^T \mathbf{q}_O) - (\mathbf{q}_O^T \mathbf{p}(r))^2}{(\mathbf{q}_O^T \mathbf{q}_O)}$$

Equ. (F)

subject to

$$0 \leq r \leq 1.$$

The results of

$$\mathbf{p}(\hat{f}), \text{ where } \hat{f}$$

satisfies Equ. (F), are the calculated bifurcation points which are saved into the nodes associated with the branching vessels in the hierarchical digraph. On the basis of the vessel hierarchy digraph, the relationships of vessel correspondence among the multiple projections are established by traversal of the associated hierarchical digraphs via the descendant and sibling arcs.

Determination of Parameters of the Biplane Imaging System

Referring now to Figs. 1, 2A-2B, 4, and step 28 of Fig. 1, the biplane imaging system includes a pair of single-plane imaging systems (Figs. 2A-2B). Each X-ray source (or focal spot) functions as the origin of 3-D coordinate space and the spatial relationship between each imaging portion of the biplane system can be characterized by a transformation in the form of a rotation matrix R and a translation vector \bar{t} . In the first projection view, let (u_i, v_i) denote the image coordinates of the i th object point, located at position (x_i, y_i, z_i) .

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1 We have $u_i = Dx_i/z_i$, $v_i = Dy_i/z_i$, where D is the perpendicular distance between the x-ray
 2 focal spot and the image plane. Let (ξ_i, η_i) denote scaled image coordinates, defined as ξ_i
 3 $= u_i/D = x_i/z_i$, $\eta_i = v_i/D = y_i/z_i$. The second projection view of the biplane imaging system
 4 can be described in terms of a second pair of image and object coordinate systems $u'v'$ and
 5 $x'y'z'$ defined in an analogous manner. Scaled image coordinates (ξ'_i, η'_i) in the second view
 6 (second projection image) for the i th object point at position (x'_i, y'_i, z'_i) are given by $\xi'_i =$
 7 $u'_i/D' = x'_i/z'_i$, $\eta'_i = v'_i/D' = y'_i/z'_i$. The geometrical relationship between the two views
 8 can be characterized by

$$\begin{bmatrix} x'_i \\ y'_i \\ z'_i \end{bmatrix} = R \cdot \left\{ \begin{bmatrix} x_i \\ y_i \\ z_i \end{bmatrix} - \vec{t} \right\} = \begin{bmatrix} r_{11} & r_{12} & r_{13} \\ r_{21} & r_{22} & r_{23} \\ r_{31} & r_{32} & r_{33} \end{bmatrix} \cdot \begin{bmatrix} x_i - t_x \\ y_i - t_y \\ z_i - t_z \end{bmatrix}.$$

9

Equ. (1)

10

11 Fig. 4 illustrates the graphical representation of the biplane system defined by a
 12 mathematical model. In the inventive method, the required prior information (*i.e.*, the
 13 intrinsic parameters of each single-plane imaging system) for determination of biplane
 14 imaging geometry includes: (1) the distance between each focal spot and its image plane, SID
 15 (focal-spot to imaging-plane distance), (2) the pixel size, p_{size} (*e.g.*, .3 mm/pixel), (3) the
 16 distance

$$\overline{ff'}$$

17

18 between the two focal spots or the known 3-D distance between two points in the projection
 19 images, and (4) for each view, an approximation of the factor MF (*e.g.*, 1.2), which is the

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1 ratio of the SID and the approximate distance of the object to the focal spot. Item (4),
2 immediately above, is optional but may provide a more accurate estimate if it is available.

3 An essential step in feature-based 3-D reconstruction from two views relies on the
4 accurate establishment of correspondence in image features, such as points or curve segments
5 between projections, as is illustrated in step 32 of Fig. 1. The bifurcation points on the
6 vascular tree are prominent features and can often be recognized in both images to facilitate
7 the determination of biplane imaging geometry. Because the vessel correspondences are
8 maintained based on the hierarchical digraphs, the correspondences of bifurcation points are
9 inherently established and can be retrieved by traversing the associated hierarchical digraphs
10 (data structures). The established pairs of bifurcation points are used for the calculation of
11 the biplane imaging geometry. Note that the "pincushion distortions" on bifurcation points
12 and image points are corrected first before the estimation of biplane imaging geometry
13 proceeds. The correction of pincushion error can be implemented based on known
14 algorithms. For example, a method described in a publication entitled "Correction of Image
15 Deformation from Lens Distortion Using Bezier Patches", Computer Vision. Graphics
16 Image Processing, Vol. 47, 1989, pp. 385-394, may be used, as is known in the art. In the
17 present inventive method, the pincushion distortion does not considerably affect the accuracy
18 of the 3-D reconstruction due to the small field of view (i.e., 100 cm SID and 17 cm x 17
19 cm II). The prior information (SID, p_{size} , MF) and the 2-D inputs are employed to serve as
20 constraints such that the intermediate solutions resulting from each iterative calculation
21 remains in the vicinity of the true solution space and converges to an "optimal" solution.

22 Initial Estimates of Biplane Imaging Geometry

23 When the input data error of corresponding points is moderate (e.g., less than 1 pixel
24 $\approx .3$ mm RMS error in coronary angiography), the estimate of the 3-D imaging geometry

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provided by the linear algorithm is generally sufficient to ensure proper convergence for further optimization. Such a linear algorithm is described in a publication by C. E. Metz and L. E. Fencil entitled "Determination of three-dimensional structure in biplane radiography without prior knowledge of the relationship between the two views: Theory," *Medical Physics*, 16 (1), pp. 45-51, Jan/Feb 1989, as is known in the art.

However, when input data error is large, the initial estimate provided by the linear algorithm may be grossly inaccurate, and the minimization procedure may become trapped in a local minimum. In the problem of biplane angiography, the centroid of a target object or the region of interest to be imaged is usually aligned with the isocenter of the imaging system as closely as possible such that the content of projection image includes the desired focus of attention at any viewing angle. The isocenter is the location between the focal spot and image intensifier with respect to the rotary motion of the gantry arm, as illustrated in Figs. 2A-2B. It is usually measured as the relative distance from the focal spot. Hence, the information with respect to the isocenter is employed and converted to the approximate MF value if the distance between the object and focal spot is not available.

The required initial estimates include a rotation matrix

$$\bar{R}$$

a unit translation vector

$$\vec{t}_0,$$

and scaled 3-D points

$$\vec{p}'_i = (\bar{x}'_i, \bar{y}'_i, \bar{z}'_i), i=1, 2, \dots, n.$$

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With large amounts of noise on the input of the 2-D corresponding points extracted from the biplane images, the estimated imaging geometry, as well as the 3-D objects by use of the linear algorithm may considerably deviate from the real solution and, therefore are not suitable to serve as the initial estimate for the refinement process. Such a situation can be identified if (1) not all of the calculated 3-D points are in front of both (or all) focal spots, (2) the RMS image point errors are large (e.g., > 50 pixels) or (3) the projections of the calculated 3-D points are not in the image plane. To remedy this problem, the estimates of

$$\bar{\mathbf{R}}, \bar{\mathbf{t}}_{\bar{\mathbf{u}}} \text{ and } \bar{\mathbf{p}}_i \text{'s}$$

must be redefined so that their values are in compliance with the initial biplane geometry set-up for the optimization. Without loss of generality, the initial estimates of the z and z' axes of the two imaging systems are taken to be orthogonal in this situation, and the unit translation vector is set to be on the x - z plane. Let (α, α') and (D, D') , denote the MF factors and SID of the biplane imaging systems in xyz and $x'y'z'$ coordinates, respectively. Two different initial solutions

$$(\bar{\mathbf{R}}_1, \bar{\mathbf{t}}_{\bar{\mathbf{u}}_1}) \text{ and } (\bar{\mathbf{R}}_2, \bar{\mathbf{t}}_{\bar{\mathbf{u}}_2})$$

are employed as follows:

$$\bar{\mathbf{R}}_1 = \begin{bmatrix} 0 & 0 & -1 \\ 0 & 1 & 0 \\ 1 & 0 & 0 \end{bmatrix}, \quad \bar{\mathbf{t}}_{\bar{\mathbf{u}}_1} = \begin{bmatrix} -\frac{D'}{\alpha' \cdot t_d} \\ 0 \\ \frac{D}{\alpha \cdot t_d} \end{bmatrix}, \text{ and } \bar{\mathbf{R}}_2 = \begin{bmatrix} 0 & 0 & 1 \\ 0 & 1 & 0 \\ -1 & 0 & 0 \end{bmatrix}, \quad \bar{\mathbf{t}}_{\bar{\mathbf{u}}_2} = \begin{bmatrix} \frac{D'}{\alpha' \cdot t_d} \\ 0 \\ \frac{D}{\alpha \cdot t_d} \end{bmatrix},$$

Equ. (2)

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where t_d represents the magnitude of \bar{t} . If the magnitude of \bar{t} is not available, an approximated measurement is calculated as follows:

$$t_d = \sqrt{\left(\frac{D}{\alpha}\right)^2 + \left(\frac{D'}{\alpha'}\right)^2}.$$

Referring now to Fig. 5, Fig. 5 illustrates the graphical representation of the predefined initial solutions. The scaled 3-D points $(\bar{x}'_i, \bar{y}'_i, \bar{z}'_i)$ defined in the $x'y'z'$ coordinate system are initialized as

$$\bar{x}'_i = \frac{u'_i}{\alpha' \cdot t_d}, \quad \bar{y}'_i = \frac{v'_i}{\alpha' \cdot t_d}, \quad \bar{z}'_i = \frac{D'}{\alpha' \cdot t_d}, \quad i=1, \dots, n,$$

where (u'_i, v'_i) denotes the 2-D input points on image plane in the $x'y'z'$ single-plane system.

Final Estimates Based on Constrained Optimization

Although the linear algorithm discussed above is computationally fast, the solution is not optimal in the presence of very noisy data (e.g., RMS error > 1 pixel). Hence, it is potentially advantageous to employ another method aiming at global optimization to improve the accuracy of the solution in image locations of corresponding points. In the approach described herein, an objective function defined as the sum of squares of the Euclidean distances between the 2-D input data and the projections of the calculated 3-D data points is employed. Given the set of 2-D points extracted from the biplane images, an "optimal" estimate of the biplane imaging geometry and 3-D object structures is obtained by minimizing:

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$$\min_{P, P'} F_1(P, P') = \sum_{i=1}^n \left\{ \left(\xi_i - \frac{x_i}{z_i} \right)^2 + \left(\eta_i - \frac{y_i}{z_i} \right)^2 + \left(\xi'_i - \frac{x'_i}{z'_i} \right)^2 + \left(\eta'_i - \frac{y'_i}{z'_i} \right)^2 \right\},$$

Equ. (3)

where n denotes the number of pairs of corresponding points extracted from the biplane images, and P and P' denote the sets of 3-D object position vectors $\bar{p}_i = (x_i, y_i, z_i)$ and $\bar{p}'_i = (x'_i, y'_i, z'_i)$, where $i=1, \dots, n$, respectively. The first two terms of the objective function $F_1(P, P')$ denote the square of distance between the input of image data and the projection of calculated 3-D data at the i th point. The last two terms are similarly defined as the square of 2-D distance error in the second image plane. Since the relationship between the two imaging systems can be characterized by a rotation matrix R and a translation vector $\bar{t} = [t_x, t_y, t_z]'$, as shown in Eq. (1), Eq. (3) can be expressed as

$$\min_{R, \bar{t}, P'} F_2(R, \bar{t}, P') = \sum_{i=1}^n \left\{ \left(\xi'_i - \frac{x'_i}{z'_i} \right)^2 + \left(\eta'_i - \frac{y'_i}{z'_i} \right)^2 + \left(\xi_i - \frac{\bar{c}_1 \cdot \bar{p}'_i + t_x}{\bar{c}_3 \cdot \bar{p}'_i + t_z} \right)^2 + \left(\eta_i - \frac{\bar{c}_2 \cdot \bar{p}'_i + t_y}{\bar{c}_3 \cdot \bar{p}'_i + t_z} \right)^2 \right\},$$

Equ. (4)

where \bar{c}_k denotes the respective k th column vectors of matrix R . From a pair of projections, the 3-D objects can only be recovered up to a scale factor, at best. This fact is reflected by the inspection of each quotient term involving the 3-D points in Equ. (4) as follows:

$$\begin{aligned} \min_{R, \bar{t}, \hat{P}'} F_3(R, \bar{t}, \hat{P}') &= \sum_{i=1}^n \left\{ \left(\xi'_i - \frac{x'_i / |\bar{t}|}{z'_i / |\bar{t}|} \right)^2 + \left(\eta'_i - \frac{y'_i / |\bar{t}|}{z'_i / |\bar{t}|} \right)^2 \right. \\ &\quad \left. + \left(\xi_i - \frac{(\bar{c}_1 \cdot \bar{p}'_i + t_x) / |\bar{t}|}{(\bar{c}_3 \cdot \bar{p}'_i + t_z) / |\bar{t}|} \right)^2 + \left(\eta_i - \frac{(\bar{c}_2 \cdot \bar{p}'_i + t_y) / |\bar{t}|}{(\bar{c}_3 \cdot \bar{p}'_i + t_z) / |\bar{t}|} \right)^2 \right\} \\ &= \sum_{i=1}^n \left\{ \left(\xi'_i - \frac{x'_i}{z'_i} \right)^2 + \left(\eta'_i - \frac{y'_i}{z'_i} \right)^2 + \left(\xi_i - \frac{\bar{c}_1 \cdot \bar{p}'_i + t_{u_x}}{\bar{c}_3 \cdot \bar{p}'_i + t_{u_z}} \right)^2 + \left(\eta_i - \frac{\bar{c}_2 \cdot \bar{p}'_i + t_{u_y}}{\bar{c}_3 \cdot \bar{p}'_i + t_{u_z}} \right)^2 \right\} \end{aligned}$$

Equ. (5)

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where

$$\hat{P}'$$

denotes the set of scaled 3-D points

$$\vec{p}'_i = (\hat{x}'_i, \hat{y}'_i, \hat{z}'_i)$$

where $i = 1, \dots, n$, to within a scale factor of the magnitude of the translation vector

$$|\vec{t}|$$

and where

$$[t_{u_x}, t_{u_y}, t_{u_z}]^T$$

denotes the unit translation vector corresponding to \vec{t} .

It is well known that any 3-D rigid motion can be uniquely decomposed into a translation and a rotation by an angle θ around an axis \vec{v}_u passing through the origin of the coordinate system. In the present inventive method, a quaternion denoted as

$$\tilde{q} = (s, \tilde{w}) = (s, w_1, w_2, w_3)$$

of norm equal to 1 is employed to represent the rotation transformation as

$$\tilde{w} = \sin(\theta/2) \vec{v}_u, \quad s = \cos(\theta/2).$$

Equ. (6)

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1 Similarly, for any quaternion

$$\tilde{\mathbf{q}} = (s, \tilde{\mathbf{w}}) = (s, w_1, w_2, w_3)$$

2

3 of norm 1, there exists a rotation R satisfying Eq. (6) and is defined as follows:

$$2 \begin{bmatrix} s^2 + (w_1)^2 - \frac{1}{2} & w_1 w_2 - s w_3 & s w_2 + w_1 w_3 \\ w_1 w_2 + s w_3 & s^2 + (w_2)^2 - \frac{1}{2} & w_2 w_3 - s w_1 \\ w_1 w_3 - s w_2 & s w_1 + w_2 w_3 & s^2 + (w_3)^2 - \frac{1}{2} \end{bmatrix}$$

4

Equ. (7)

5 With this quaternion representation, Eq. (4) can be rewritten as:

$$\min_{\tilde{\mathbf{q}}, \tilde{\mathbf{r}}, \tilde{\mathbf{p}}} F_4(\tilde{\mathbf{q}}, \tilde{\mathbf{r}}, \tilde{\mathbf{p}}) = \sum_{i=1}^n \left[\left(\xi'_i - \frac{\tilde{x}'_i}{\tilde{z}'_i} \right)^2 + \left(\eta'_i - \frac{\tilde{y}'_i}{\tilde{z}'_i} \right)^2 + \left\{ \begin{aligned} & \xi'_i - \frac{2(s^2 + w_1^2 + 1/2)\tilde{x}'_i + 2(w_1 w_2 + s w_3)\tilde{y}'_i + 2(w_1 w_3 - s w_2)\tilde{z}'_i + t_{a_i}}{2(s w_2 + w_1 w_3)\tilde{x}'_i + 2(w_2 w_3 - s w_1)\tilde{y}'_i + 2(s^2 + w_3^2 - \frac{1}{2})\tilde{z}'_i + t_{a_i}} \right\}^2 \\ & + \left\{ \begin{aligned} & \eta'_i - \frac{2(w_1 w_2 - s w_3)\tilde{x}'_i + 2(s^2 + w_2^2 - 1/2)\tilde{y}'_i + 2(s w_1 + w_2 w_3)\tilde{z}'_i + t_{b_i}}{2(s w_2 + w_1 w_3)\tilde{x}'_i + 2(w_2 w_3 - s w_1)\tilde{y}'_i + 2(s^2 + w_3^2 - \frac{1}{2})\tilde{z}'_i + t_{a_i}} \right\}^2 \end{aligned} \right] \end{aligned}$$

Equ. (8)

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1 subject to the constraints:

$$C_1: s^2 + (w_1)^2 + (w_2)^2 + (w_3)^2 = 1$$

$$C_2: (t_{u_i})^2 + (t_{v_i})^2 + (t_{w_i})^2 = 1$$

$$C_3: 0 < \hat{z}_i, i = 1, \dots, n$$

$$C_4: 0 < 2(sw_2 + w_1 w_3)\hat{x}_i + 2(w_2 w_3 - sw_1)\hat{y}_i + 2(s^2 + (w_3)^2 - \frac{1}{2})\hat{z}_i + t_{u_i} = \hat{z}_i, i = 1, \dots, n,$$

2 where constraint C_1 characterizes the quaternion norm, constraint C_2 ensures a unit
3 translation vector, and constraints C_3 and C_4 force the scaled coordinates \hat{z}'_i and \hat{z}_i to be in
4 front of the respective focal spots.

5 If the isocenter distances of employed biplane imaging systems or MF factors are
6 available, the constraints C_3 and C_4 in Eq.(8) can be modified as:

$$C_3: \frac{d' - \delta h}{|\vec{t}|} < \hat{z}_i < \frac{d' + \delta h}{|\vec{t}|}, i = 1, \dots, n$$

$$C_4: \frac{d - \delta h}{|\vec{t}|} < \hat{z}_i \approx 2[(sw_2 + w_1 w_3)\hat{x}_i + (w_2 w_3 - sw_1)\hat{y}_i + (s^2 + (w_3)^2 - \frac{1}{2})\hat{z}_i] + t_{u_i} < \frac{d + \delta h}{|\vec{t}|}$$

$i = 1, \dots, n,$

7

8 where $d = D/\alpha$ and $d' = D'/\alpha'$ are the approximate distances between the object's centroid
9 and the respective focal spots, α and α' denote the MF factors and δ_h ($\approx 12.5 \pm 2.0$ cm)
10 denotes the maximal length of the heart along a long-axis view at end-diastole, as is known
11 and described by A.E. Weyman in a publication entitled "*Cross-Sectional*
12 *Echocardiography*," Lea & Febiger, Philadelphia, 1982. For each 3-D object point, the ray
13 connecting that point and the focal spot intersects the image plane near the associated 2-D
14 image point, even when the input data is corrupted by noise. In addition to the constraints

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1 imposed on the z and z' coordinates, two other constraints are incorporated to confine the x ,
 2 x' , y , and y' coordinates of each calculated 3-D point as follows:

3 For each 3-D object point, the ray connecting that point and the focal spot intersects
 4 the image plane near the associated 2-D image point, even when the input data are corrupted
 5 by noise. In addition to the constraints imposed on the z and z' coordinates, two other
 6 constraints are incorporated to confine the x , x' , y , and y' coordinates of each calculated 3-D
 7 point as follows:

$$C_5 : \left(\frac{\hat{x}_i}{\hat{z}_i} - \xi_i \right)^2 + \left(\frac{\hat{y}_i}{\hat{z}_i} - \eta_i \right)^2 \leq \left(\frac{\delta_c}{D p_{size}} \right)^2, \quad i=1, \dots, n,$$

$$C_6 : \left(\frac{\hat{x}'_i}{\hat{z}'_i} - \xi'_i \right)^2 + \left(\frac{\hat{y}'_i}{\hat{z}'_i} - \eta'_i \right)^2 \leq \left(\frac{\delta_c}{D' p_{size}} \right)^2, \quad i=1, \dots, n,$$

8

9 where δ_c defines the radius of a circular disk (e.g., 20 pixels) centered at (ξ_i, η_i) or (ξ'_i, η'_i)
 10 and p_{size} represents the pixel size.

11 Referring now to Fig. 6, Fig. 6 shows the bounding regions based on the employed
 12 constraint C_3 to C_6 in $x'y'z'$ system. If two initial solutions are employed (as described under
 13 the subheading of Initial Estimates of Biplane Imaging Geometry), in general, two sets of
 14 biplane imaging geometry and their associated 3-D scaled object points will be obtained:

15

$$[R_1, \vec{t}_{u_1}, (\hat{x}'_1, \hat{y}'_1, \hat{z}'_1), i=1, \dots, n]$$

16

17 and

18

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$$[R_2, \vec{t}_{u_2}, (\hat{x}'_{2_i}, \hat{y}'_{2_i}, \hat{z}'_{2_i}), i=1, \dots, n].$$

1

2 Referring now to Fig. 7, Fig. 7 illustrates a typical example by use of several object
 3 point RMS errors on image points associated with the true solutions defined by one imaging
 4 geometry

$$(e.g., R_1 \text{ and } \vec{t}_{u_1})$$

5

6 is smaller than those defined by the other imaging geometry

$$(e.g., R_2 \text{ and } \vec{t}_{u_2}).$$

7

8 Therefore, the calculated imaging parameters, which have a smaller RMS error on the image
 9 points, are selected as the optimal solution. To determine the absolute size of the object, the
 10 magnitude of the translation vector (*i.e.*, the distance between the two focal spots

$$\overline{ff'},$$

11

12 or the real 3-D distance between any two object points projected onto the biplane images
 13 needs to be known. In the former case, the actual 3-D object points can be recovered easily
 14 by multiplying the scaled object points by the magnitude. Otherwise, the scale factor S_f is
 15 calculated and employed to obtain the absolute 3-D object point as

$$\begin{bmatrix} x_i \\ y_i \\ z_i \end{bmatrix} = S_f \cdot \begin{bmatrix} \hat{x}_i \\ \hat{y}_i \\ \hat{z}_i \end{bmatrix}, \text{ where } S_f = \frac{L_d}{\sqrt{(\hat{x}_{p_1} - \hat{x}_{p_2})^2 + (\hat{y}_{p_1} - \hat{y}_{p_2})^2 + (\hat{z}_{p_1} - \hat{z}_{p_2})^2}},$$

and L_d denotes the known 3-D distance associated with the two scaled 3-D object points

$$(\hat{x}_{p_1}, \hat{y}_{p_1}, \hat{z}_{p_1}) \text{ and } (\hat{x}_{p_2}, \hat{y}_{p_2}, \hat{z}_{p_2}).$$

Recovery of 3-D Spatial Information

After the biplane imaging geometry that defines the two views is obtained, the orientation information is used to establish the point correspondences on vessel centerlines in the pair of images and is further used to calculate 3-D morphologic structures of coronary arterial tree, as is illustrated in step 34 of Fig. 1. The calculated imaging geometry in conjunction with the epipolar constraints are employed as the framework for establishing the point correspondences on the vessel centerlines based on the two identified 2-D coronary arterial trees.

According to the epipolar constraints, the correspondence of a point in one image must lie on the epipolar line in the other image. Two types of ambiguity may arise in the two-view correspondence problem: (1) the epipolar line may intersect more than one vessel in the coronary arterial tree, and (2) the epipolar line may intersect more than one point on a single vessel. The first ambiguity is resolved by means of the constructed hierarchical digraph defining the anatomy of the 2-D coronary arterial tree such that the epipolar constraints are applied iteratively to every pair of corresponding vessels in the two coronary

-30-

1 trees. For example, the corresponding centerline points of the left anterior descending artery
 2 in the angiogram acquired from the first view is uniquely determine by finding the
 3 intersections of the epipolar line and the 2-D centerline of the left anterior descending artery
 4 in the angiogram acquired from the second view.

5 When the intersection point is calculated, each 2-D vessel centerline is modeled by
 6 a spline-based curve-fitting function $f(s) = (x_i, y_i)$, $0 \leq s \leq 1$ (the same method used for
 7 calculation of bifurcation points described above) where s is the parametric argument defining
 8 the location of points (x_i, y_i) on the vessel centerline. If there are n intersection points
 9 between the epipolar line and the vessel centerline due to the tortuous vessel shape, the
 10 locations of these points can be defined based on the parametric arguments (*e.g.*, $f(s_1)$,
 11 $f(s_2), \dots, f(s_n)$, $s_1 = 0.2$, $s_2 = 0.35, \dots, s_n = 0.5$). The point with the parametric argument s_k ,
 12 $1 \leq k \leq n$ is selected as the desired corresponding point if s_k is the smallest value larger
 13 than the parametric argument of the last detected corresponding point. Based on such a
 14 method, the second type of ambiguity is resolved.

15 With the point correspondences on 2-D vessel centerlines (ξ_i, η_i) and (ξ'_i, η'_i) and the
 16 imaging geometry, the 3-D centerline points of coronary arteries (x_i, y_i, z_i) 's can then be
 17 calculated based on the following equations:

$$\begin{bmatrix} r_{11}-r_{31}\xi'_i & r_{12}-r_{32}\xi'_i & r_{13}-r_{33}\xi'_i \\ r_{21}-r_{31}\eta'_i & r_{22}-r_{32}\eta'_i & r_{23}-r_{33}\eta'_i \\ 1 & 0 & -\xi_i \\ 0 & 1 & -\eta_i \end{bmatrix} \cdot \begin{bmatrix} x_i \\ y_i \\ z_i \end{bmatrix} = \begin{bmatrix} \vec{a} \cdot \vec{t} \\ \vec{b} \cdot \vec{t} \\ 0 \\ 0 \end{bmatrix},$$

Equ. (9)

-31-

1 where \vec{a} and \vec{b} are two vectors defined as follows:

$$\vec{a} = \begin{bmatrix} (r_{11} - r_{31}\xi'_i) \\ (r_{12} - r_{32}\xi'_i) \\ (r_{13} - r_{33}\xi'_i) \end{bmatrix}, \quad \vec{b} = \begin{bmatrix} (r_{21} - r_{31}\eta'_i) \\ (r_{22} - r_{32}\eta'_i) \\ (r_{23} - r_{33}\eta'_i) \end{bmatrix}$$

2 Equ. (10)

3 and r_{ij} denotes the component of the rotation matrix R .

4 Rendering of Reconstructed 3-D Coronary Tree and Estimation of an Optimal View

5 After the 3-D vessel centerlines are obtained which define the 3-D location of the
6 arterial tree, as shown in step 34 of Fig. 1, the anatomical morphology of the arterial tree
7 is generated by a surface based reproduction technique, as illustrated in step 36 of Fig. 1,
8 as is known in the art. Such a surface based reproduction technique is described by S.Y.
9 Chen, K.R. Hoffmann, C.T. Chen, and J.D. Carroll in a publication entitled "Modeling the
10 Human Heart based on Cardiac Tomography," *SPIE*, vol. 1778, 1992, pp. 14-18.

11 The 3-D lumen surface is represented by a sequence of cross-sectional contours.
12 Each contour V_i along the vessel is represented by a d_i -mm circular disk centered at and
13 perpendicular to the 3-D vessel centerline. The surface between each pair of consecutive
14 contours V_i and V_{i+1} is generated based upon a number of polygonal patches. Utilizing the
15 modeled lumen surfaces, the morphology of the reconstructed coronary arterial tree is
16 reproduced by employing the technique of computer graphics, as is known in the art.

17 When an arbitrary computer-generated image is produced, the gantry information
18 defining the current projection is calculated in the form of *LAO/RAO* (on the y - z plane) and
19 *CAUD/CRAN* (on the x - z plane) angles by which the gantry arm moves along the *LAO/RAO*

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angle followed by the CAUD/CRAN angle. The focal spot of the gantry can be formulated as

$$\begin{aligned}
 R_x(\gamma)R_y(-\beta) &= \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos(\gamma) & \sin(\gamma) \\ 0 & -\sin(\gamma) & \cos(\gamma) \end{bmatrix} \cdot \begin{bmatrix} \cos(-\beta) & 0 & -\sin(-\beta) \\ 0 & 1 & 0 \\ \sin(-\beta) & 0 & \cos(-\beta) \end{bmatrix} \\
 &= \begin{bmatrix} \cos(-\beta) & 0 & -\sin(-\beta) \\ \sin(\gamma)\sin(-\beta) & \cos(\gamma) & \sin(\gamma)\cos(-\beta) \\ \cos(\gamma)\sin(-\beta) & -\sin(\gamma) & \cos(\gamma)\cos(-\beta) \end{bmatrix}
 \end{aligned}$$

Equ. (11)

where R_x and R_y denote the rigid rotations with respect to the x -axis and y -axis, respectively, and where γ and β denote the LAO and CAUD angles, respectively.

Let p_i , $i = 0, 1, \dots, m$ denote the points on the centerline of a 3-D vessel. Let

$$\vec{l}_j = [l_{j_x}, l_{j_y}, l_{j_z}]^t \text{ and } l_j, j=1, 2, \dots, m$$

denote the vector and length of the segments between p_{j-1} and p_j , respectively. The minimal foreshortening of the vessel segments are obtained in terms of the gantry orientation (γ and β angles) by minimizing the objective function as follows:

$$\begin{aligned}
 \min_{\gamma, \beta} F(p, \gamma, \beta) &= \sum_{j=1}^m \|\vec{l}_j \cos(\theta_j)\|^2 \\
 &= \sum_{j=1}^m (\vec{l}_j \cdot \vec{z})^2,
 \end{aligned}$$

Equ. (12)

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subject to the constraints

$$-90^\circ < \gamma < 90^\circ, \quad -40^\circ < \beta < 40^\circ,$$

where "." denotes the inner product and θ_j is the angle between the directional vector \vec{T}_j and the projection vector \vec{z}_p is defined as

$$\vec{z}_p = \begin{bmatrix} -\cos(\gamma)\sin(\beta) \\ -\sin(\gamma) \\ \cos(\gamma)\cos(\beta) \end{bmatrix}.$$

Equ. (13)

In prior art methods, due to the problem of vessel overlap and vessel foreshortening, multiple projections are necessary to adequately evaluate the coronary arterial tree using arteriography. Hence, the patient may receive additional or unneeded radiation and contrast material during diagnostic and interventional procedures. This known traditional trial and error method may provide views in which overlapping and foreshortening are somewhat minimized, but only in terms of the subjective experience-based judgement of the angiographer. In the present inventive method, the reconstructed 3-D coronary arterial tree can be rotated to any selected viewing angle yielding multiple computer-generated projections to determine for each patient which standard views are useful and which are of no clinical value due to excessive overlap. Therefore, the 3-D computer assistance provides a means to improve the quality and utility of the images subsequently acquired.

Experimental Results

The accuracy of the present inventive method was evaluated by use of bifurcation points in a computer-simulated coronary arterial tree. For assessment of the rotation matrix, R is further decomposed into a rotation around an axis \vec{v}_u (a unit vector) passing through the

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origin of the coordinate system with the angle θ . The differences between the calculated and real rotation axes E_v and rotation angles E_θ are employed for error analysis. The error in the translation vector E_T is the angle between the real and calculated translation vectors. The error in the 3-D absolute position E_{3d} is defined as the RMS distance between the calculated and the real 3-D data sets; while the error in the 3-D configuration

$$E_{3d}$$

is defined as the RMS distance between the calculated and the real 3-D data sets after the centroids of these two data sets have been made to coincide. In simulated experiments, the parameters of the biplane imaging geometry were varied to investigate the effects of the system geometry on the accuracy with which the 3-D point positions could be recovered.

Both D and D' were equal to 100 cm.

To assess the reliability of the technique under realistic conditions, a set of experiments was simulated by adding independent errors to the 2-D vessel centerlines resulting from the projection of the simulated 3-D arterial tree. The effect of the relative angle between the biplane imaging views was assessed by varying ϕ from 30° to 150° . By use of the computer simulated coronary arterial tree, RMS errors in angles defining the R matrix and \bar{t} vector were less than 0.5 (E_v), 1.2 (E_θ), and 0.7 (E_T) degrees, respectively, when ten corresponding points were used with RMS normally distributed errors varying from 0.7 - 4.2 pixels (0.21 - 1.32 mm) in fifty configurations; when only the linear based Metz-Fencil method was employed, the respective errors varied from 0.5 - 8.0 degrees, 6.0 - 40.0 degrees, and 3.7 - 34.1 degrees. The simulation shows substantial improvement in the estimation of biplane imaging geometry based on the new technique, which facilitates

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1 accurate reconstruction of 3-D coronary arterial structures. The RMS errors in 3-D absolute
2 position (E_{3d}) and configuration

$$(E_{3d})$$

3 of the reconstructed arterial tree were 0.9 - 5.5 mm and 0.7 - 1.0 mm, respectively. The
4 following table shows one of the simulation results based on an orthogonal biplane set-up:

RMS error in 2-D centerlines (pixel)	Error in imaging parameters				RMS error in 3-D	
	$E_{\bar{v}}$	E_{θ}	$E_{\bar{\tau}}$	E_{shift}	$E_{\bar{\tau}}$	E_{3d}
0.5	0.30°	0.61°	0.11°	1.8mm	1.08mm	0.9mm
1.0	0.33°	0.72°	0.40°	6.4mm	1.37mm	3.7mm
1.5	0.49°	1.19°	0.64°	10.2 mm	1.56mm	5.5mm
2.0	0.40°	0.65°	0.64°	6.5mm	1.79mm	3.6mm
2.5	0.42°	0.95°	0.40°	6.4mm	0.74mm	2.8mm
3.0	0.42°	0.86°	0.40°	6.5mm	0.96mm	2.3mm

15 where

$$E_{\bar{v}}$$

16 denotes the deviation angle between the true \bar{v} and the calculated \bar{v}' rotational axes and E_{θ}

17 denotes the angle difference between the true θ and the calculated θ' rotational angles.

18 Note that the 3-D absolute position error is due primarily to displacement error $E_{shift} =$
19 $(D_{f-f'} \cdot E_{\bar{\tau}})$ that results from inaccurate estimation of the translation vector, where $D_{f-f'}$ is the
20 distance between the focal spots of two imaging systems and $E_{\bar{\tau}}$ denotes the deviation angle
21 between the real and calculated translation vectors. The RMS error in the 3-D configuration

$$(E_{3d})$$

22 decreases due to the reduction of the displacement error after the centroids of the real and
23 calculated data are made to coincide. In general, the results show a great similarity between

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1 the reconstructed and the real 3-D vessel centerlines. The simulation shows highly accurate
2 results in the estimation of biplane imaging geometry, vessel correspondences (less than 2 mm
3 RMS error), and 3-D coronary arterial structures (less than 2 mm RMS error in configuration
4 and 0.5 cm RMS error in absolute position, respectively) when a computer-simulated coronary
5 arterial tree is used.

6 Angiograms of fifteen patients were analyzed where each patient had multiple biplane
7 image acquisitions. The biplane imaging geometry was first determined without the need of
8 a calibration object, and the 3-D coronary arterial trees including the left and the right coronary
9 artery systems were reconstructed. Similarity between the real and reconstructed arterial
10 structures was excellent.

11 Conclusions

12 The present inventive method is novel in several ways: (1) the 3-D coronary vasculature
13 is reconstructed from a pair of projection angiograms based on a biplane imaging system or
14 multiple pairs of angiograms acquired from a single-plane system in the same phase of the
15 cardiac cycle at different viewing angles without use of a calibration object to achieve
16 accuracies in magnification and imaging geometry of better than 2% and three degrees,
17 respectively; (2) a beating 3-D coronary vasculature can be reproduced throughout the cardiac
18 cycles in the temporal sequences of images to facilitate the study of heart movement; (3) the
19 choice of an optimal view of the vasculature of interest can be achieved on the basis of the
20 capability of rotating the reconstructed 3-D coronary arterial tree; and (4) the inventive method
21 can be implemented on most digital single-plane or biplane systems. A calculated 3-D coronary
22 tree for each patient predicts which projections are clinically useful thus providing an optimal
23 visualization strategy which leads to more efficient and successful diagnostic and therapeutic

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1 procedures. The elimination of coronary artery views with excessive overlap may reduce
2 contrast and radiation.

3 Note that the present inventive method is not limited to X-ray based imaging systems.
4 For example, suitable imaging systems may include particle-beam imaging systems, radar
5 imaging systems, ultrasound imaging systems, photographic imaging systems, and laser imaging
6 systems. Such imaging systems are suitable when perspective-projection images of the target
7 object are provided by the systems.

8 Please refer to Appendix A for a source code listing of the above-described method.
9 The software is written in C Programming Language including GL Graphics Library Functions
10 and Tk. Tcl Library functions compiled on a Unix-based C Compiler.

11 Specific embodiments of a method and apparatus for three-dimensional reconstruction
12 of coronary vessels from angiographic images according to the present invention have been
13 described for the purpose of illustrating the manner in which the invention may be made and
14 used. It should be understood that implementation of other variations and modifications of the
15 invention and its various aspects will be apparent to those skilled in the art, and that the
16 invention is not limited by the specific embodiment[s] described. It is therefore contemplated
17 to cover by the present invention any and all modifications, variations, or equivalents that fall
18 within the true spirit and scope of the basic underlying principles disclosed and claimed herein.

CLAIMS

WHAT IS CLAIMED IS:

1 1. A method for three-dimensional reconstruction of a target object from two-
2 dimensional images, said target object having a plurality of branch-like vessels, the method
3 comprising the steps of:

4 a) placing the target object in a position to be scanned by an imaging system, said
5 imaging system having a plurality of imaging portions;

6 b) acquiring a plurality of projection images of the target object, each imaging portion
7 providing a projection image of the target object, each imaging portion disposed at an unknown
8 orientation relative to the other imaging portions;

9 c) identifying two-dimensional vessel centerlines for a predetermined number of the
10 vessels in each of the projection images;

11 d) creating a vessel hierarchy data structure for each projection image, said data
12 structure including the identified two-dimensional vessel centerlines defined by a plurality of
13 data points in the vessel hierarchy data structure;

14 e) calculating a predetermined number of bifurcation points for each projection image
15 by traversing the corresponding vessel hierarchy data structure, said bifurcation points defined
16 by intersections of the two-dimensional vessel centerlines;

17 f) determining a transformation in the form of a rotation matrix and a translation vector
18 utilizing the bifurcation points corresponding to each of the projections images, said rotation
19 matrix and said translation vector representing imaging parameters corresponding to the
20 orientation of each imaging portion relative to the other imaging portions of the imaging
21 system;

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1 g) utilizing the data points and the transformation to establish a correspondence between
2 the two-dimensional vessel centerlines corresponding to each of the projection images such that
3 each data point corresponding to one projection image is linked to a data point corresponding
4 to the other projection images, said linked data points representing an identical location in the
5 vessel of the target object;

6 h) calculating three-dimensional vessel centerlines utilizing the two-dimensional vessel
7 centerlines and the correspondence between the data points of the two-dimensional vessel
8 centerlines; and

9 i) reconstructing a three-dimensional visual representation of the target object based on
10 the three-dimensional vessel centerlines and diameters of each vessel estimated along the three-
11 dimensional centerline of each vessel.

12 2. The method of claim 1 wherein the two-dimensional images are two-dimensional
13 angiographic projection images.

14 3. The method of claim 1 wherein the target object is scanned substantially
15 simultaneously by the plurality of imaging portions of the imaging system.

16 4. The method of claim 3 wherein a moving target object experiences substantially
17 no movement during said substantially simultaneous scanning such that the projection images
18 provided by the plurality of imaging portions represent images of the target object acquired
19 substantially at a same point in time.

20 5. The method of claim 1 wherein the imaging system is a biplane imaging system
21 having two imaging portions, said imaging portions configured to simultaneously scan the target
22 object and provide biplane projection images of the target object.

23 6. The method of claim 1 wherein the imaging system is an X-ray based projection
24 imaging system.

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1 7. The method of claim 1 wherein the imaging system is selected from the group
2 of non-orthogonal imaging systems consisting of X-ray imaging systems, particle-beam imaging
3 systems, radar imaging systems, ultrasound imaging systems, photographic imaging systems,
4 and laser imaging systems.

5 8. The method of claim 1 wherein the step of identifying the two-dimensional vessel
6 centerlines includes the steps of determining a maximum vessel diameter at a beginning portion
7 of the vessel and determining a minimum vessel diameter at an ending portion of the vessel.

8 9. The method of claim 1 wherein the step of identifying the two-dimensional vessel
9 centerlines is performed by a human operator.

10 10. The method of claim 1 wherein the step of identifying the two-dimensional vessel
11 centerlines includes the step of identifying at least six vessels in each projection image.

12 11. The method of claim 10 wherein the step of identifying at least six vessels in
13 each projection image permits five bifurcation points to be calculated.

14 12. The method of claim 1 wherein the predetermined number of bifurcation points
15 calculated for each projection image is at least five bifurcation points.

16 13. The method of claim 1 wherein the step of reconstructing the visual
17 representation of the target object includes the steps of modeling the target object by calculating
18 estimated vessel diameters along the three-dimensional centerline of each vessel based on the
19 minimum and maximum diameter and a predetermined change in diameter per unit length along
20 the vessel.

21 14. The method of claim 1 wherein the step of reconstructing a visual representation
22 of the target object further includes the steps of:
23

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1 a) providing an optimal three-dimensional visual representation of the target object such
2 that vessel overlap and vessel foreshortening are minimized in the visual representation by
3 rotating the three-dimensional visual representation in at least one of three dimensions;

4 b) calculating image parameters corresponding to the rotated three-dimensional visual
5 representation; and

6 c) providing said calculated parameters to the imaging system to permit the imaging
7 system to further scan the target object such that optimal projection images of the target object
8 are produced.

9 15. A method for three-dimensional reconstruction of a target object from two-
10 dimensional images, said target object having a plurality of branch-like vessels, the method
11 comprising the steps of:

12 a) placing the target object in a position to be scanned by a biplane imaging system, said
13 biplane imaging system having first and second imaging portions;

14 b) acquiring biplane projection images of the target object, each imaging portion
15 providing a biplane projection image of the target object, each imaging portion disposed at an
16 unknown orientation relative to the other imaging portion;

17 c) identifying two-dimensional vessel centerlines for a predetermined number of the
18 vessels in each of the biplane projection images;

19 d) creating a vessel hierarchy data structure for each biplane projection image, said data
20 structure including the identified two-dimensional vessel centerlines defined by a plurality of
21 data points in the vessel hierarchy data structure;

22 e) calculating a predetermined number of bifurcation points for each biplane projection
23 image by traversing the corresponding vessel hierarchy data structure, said bifurcation points
24 defined by intersections of the two-dimensional vessel centerlines;

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1 f) determining a transformation in the form of a rotation matrix and a translation vector
2 utilizing the bifurcation points corresponding to each of the biplane projections images, said
3 rotation matrix and said translation vector representing biplane imaging parameters
4 corresponding to the orientation of each imaging portion relative to the other imaging portion
5 of the biplane imaging system;

6 g) utilizing the data points and the transformation to establish a correspondence between
7 the two-dimensional vessel centerlines corresponding to each of the biplane projection images
8 such that each data point corresponding to one biplane projection image is linked to a data point
9 corresponding to the other biplane projection image, said linked data points representing an
10 identical location in the vessel of the target object;

11 h) calculating three-dimensional vessel centerlines utilizing the two-dimensional vessel
12 centerlines and the correspondence between the data points of the two-dimensional vessel
13 centerlines; and

14 i) reconstructing a three-dimensional visual representation of the target object based on
15 the three-dimensional vessel centerlines and diameters of each vessel estimated along the three-
16 dimensional centerline of each vessel.

17 16. The method of claim 15 wherein the two-dimensional images are two-dimensional
18 angiographic projection images.

19 17. The method of claim 15 wherein the target object is scanned substantially
20 simultaneously by the plurality of imaging portions of the biplane imaging system.

21 18. The method of claim 17 wherein a moving target object experiences substantially
22 no movement during said substantially simultaneous scanning such that the projection images
23 provided by the plurality of imaging portions represent images of the target object acquired
24 substantially at a same point in time.

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1 19. The method of claim 15 wherein the step of reconstructing a visual representation
2 of the target object further includes the steps of:

3 a) providing an optimal three-dimensional visual representation of the target object such
4 that vessel overlap and vessel foreshortening are minimized in the visual representation by
5 rotating the three-dimensional visual representation in at least one of three dimensions;

6 b) calculating image parameters corresponding to the rotated three-dimensional visual
7 representation; and

8 c) providing said calculated parameters to the biplane imaging system to permit the
9 imaging system to further scan the target object such that optimal projection images of the
10 target object are produced.

11 20. A method for three-dimensional reconstruction of a target object from two-
12 dimensional images, said target object having a plurality of branch-like vessels, the method
13 comprising the steps of:

14 a) placing the target object in a position to be scanned by a single-plane system, said
15 imaging system having one imaging portion;

16 b) acquiring projection images of the target object, the imaging portion providing a
17 plurality of projection images of the target object produced at different times, each projection
18 image produced during an identical phase of a cardiac cycle of the target object, the imaging
19 portion corresponding to one of the plurality of projection images disposed at an unknown
20 orientation relative to the imaging portion corresponding to the others of the plurality of
21 projection images;

22 c) identifying two-dimensional vessel centerlines for a predetermined number of the
23 vessels in each of the projection images;

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1 d) creating a vessel hierarchy data structure for each projection image, said data
2 structure including the identified two-dimensional vessel centerlines defined by a plurality of
3 data points in the vessel hierarchy data structure;

4 e) calculating a predetermined number of bifurcation points for each projection image
5 by traversing the corresponding vessel hierarchy data structure, said bifurcation points defined
6 by intersections of the two-dimensional vessel centerlines;

7 f) determining a transformation in the form of a rotation matrix and a translation vector
8 utilizing the bifurcation points corresponding to each of the projections images, said rotation
9 matrix and said translation vector representing imaging parameters corresponding to the relative
10 orientations of the imaging portion corresponding to the plurality of projection images;

11 g) utilizing the data points and the transformation to establish a correspondence between
12 the two-dimensional vessel centerlines corresponding to each of the projection images such that
13 each data point corresponding to one projection image is linked to a data point corresponding
14 to the other projection images, said linked data points representing an identical location in the
15 vessel of the target object;

16 h) calculating three-dimensional vessel centerlines utilizing the two-dimensional vessel
17 centerlines and the correspondence between the data points of the two-dimensional vessel
18 centerlines; and

19 i) reconstructing a three-dimensional visual representation of the target object based on
20 the three-dimensional vessel centerlines and diameters of each vessel estimated along the three-
21 dimensional centerline of each vessel.

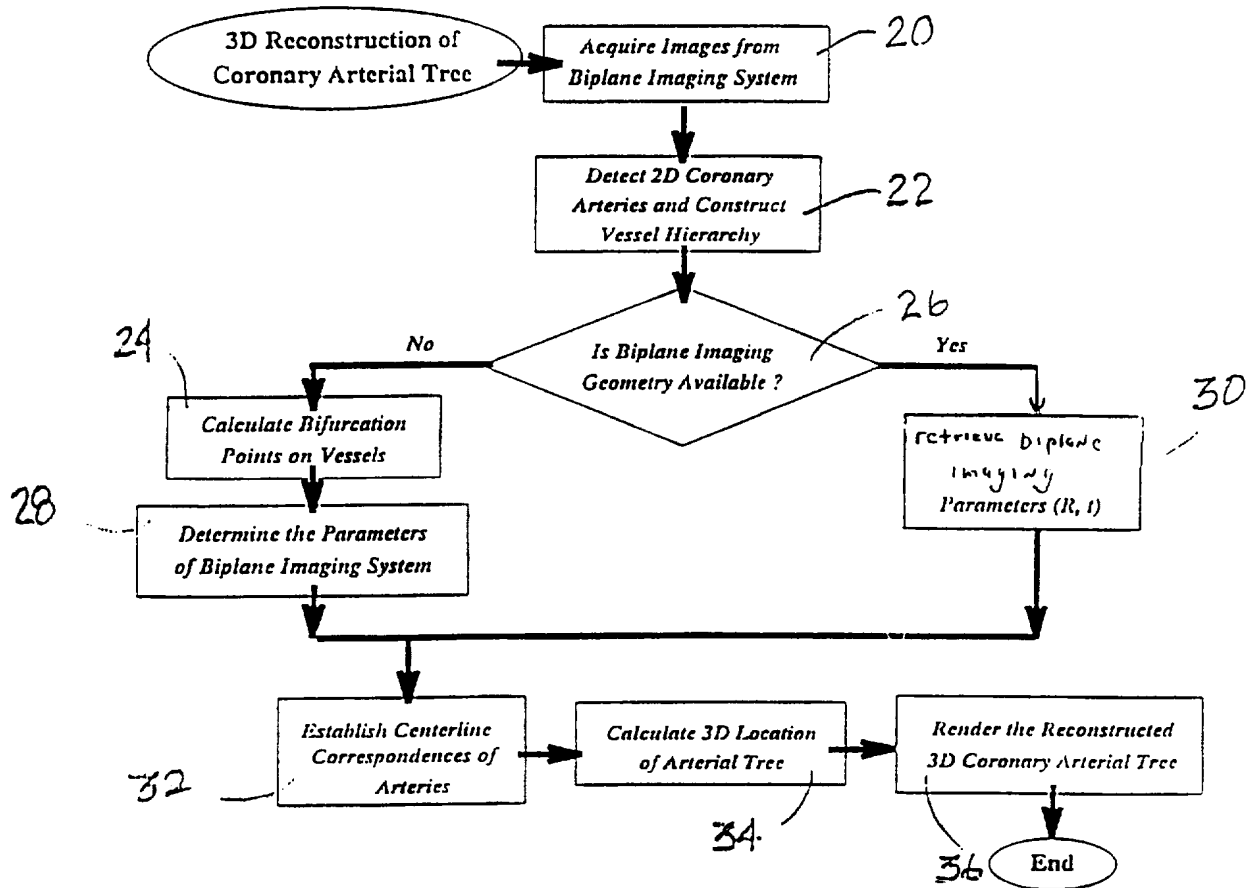
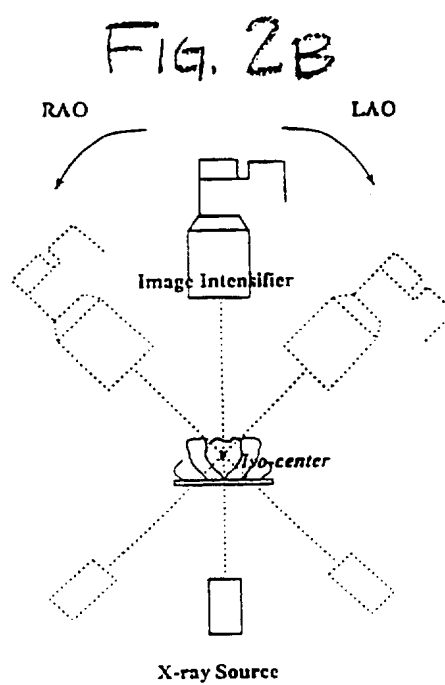
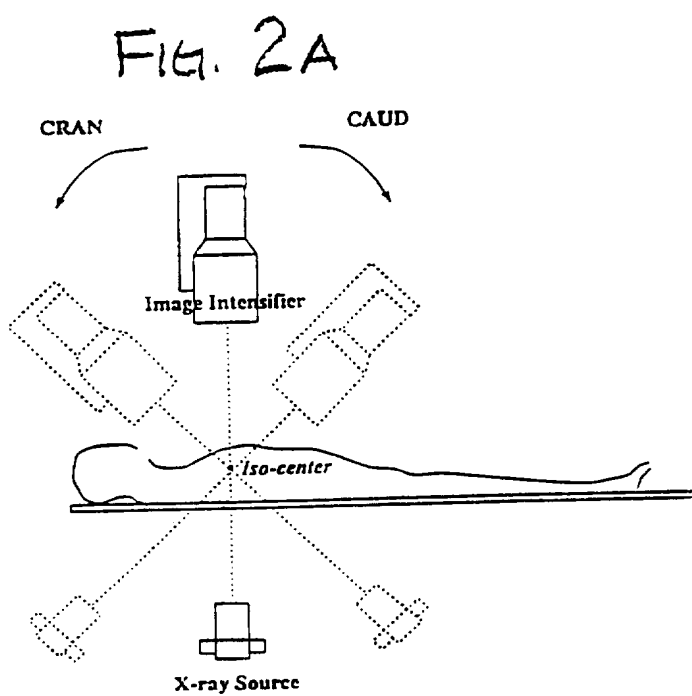


FIG. 1



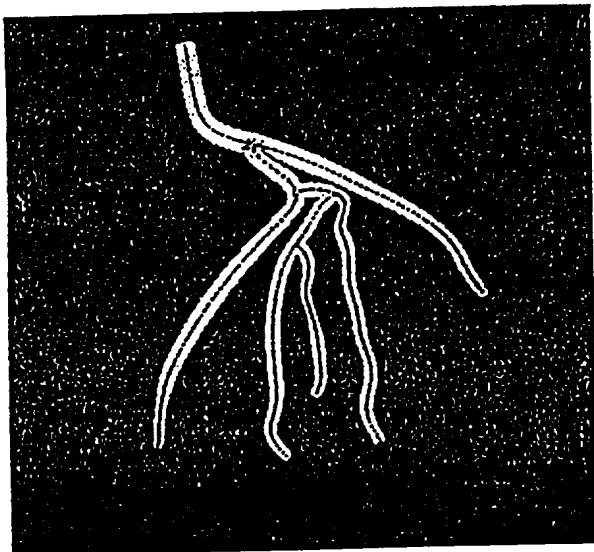


FIG. 3A

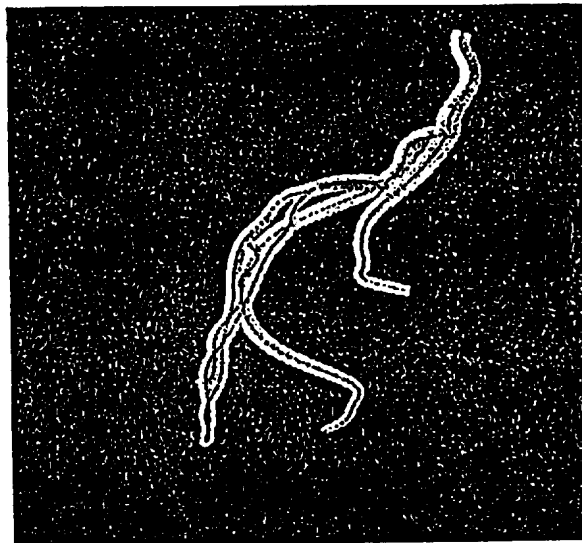


FIG. 3B

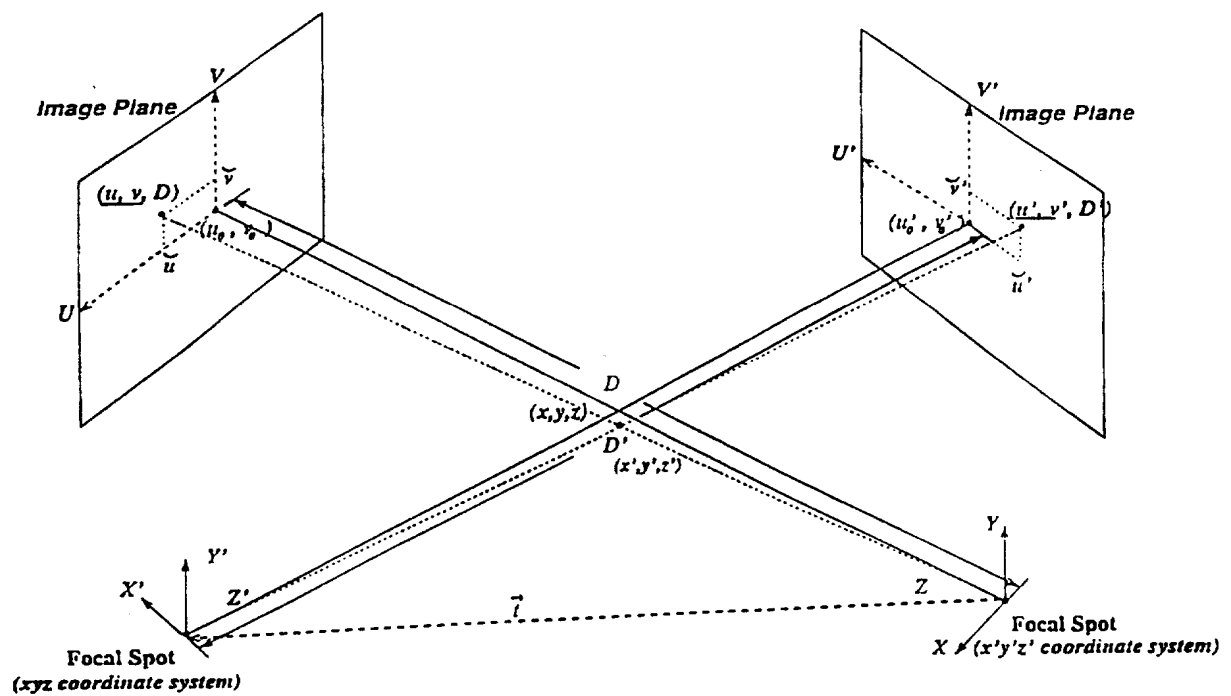


FIG. 4

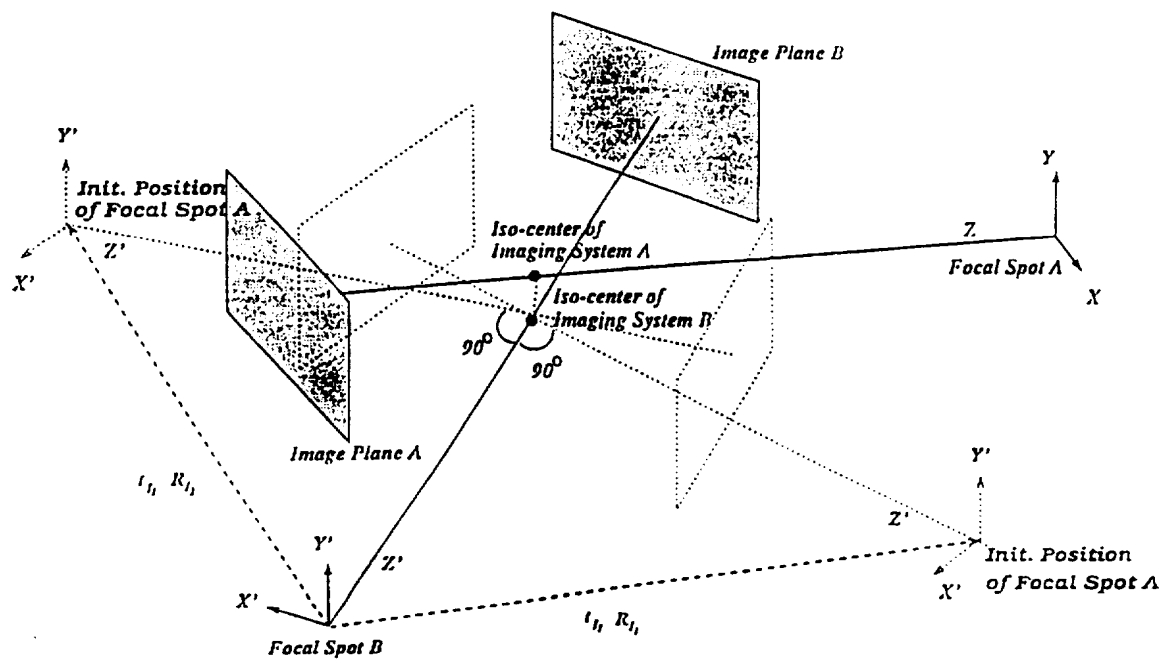


FIG. 5

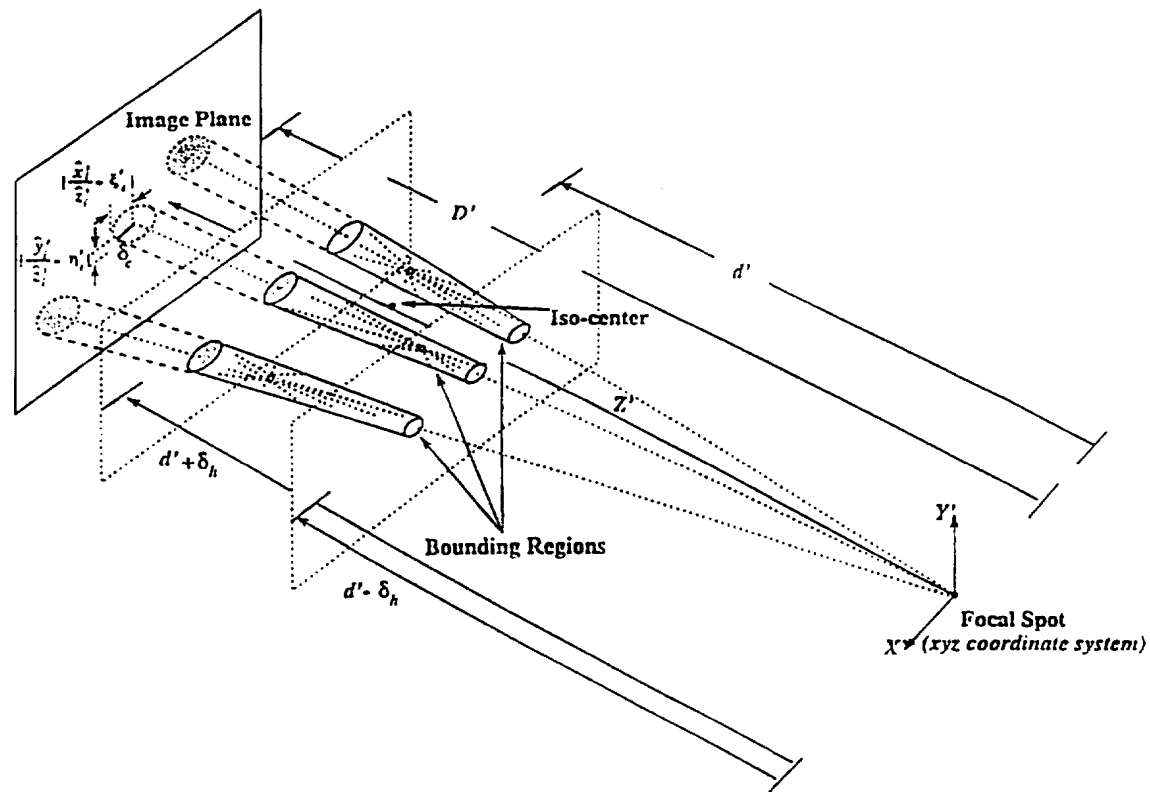


FIG. 6

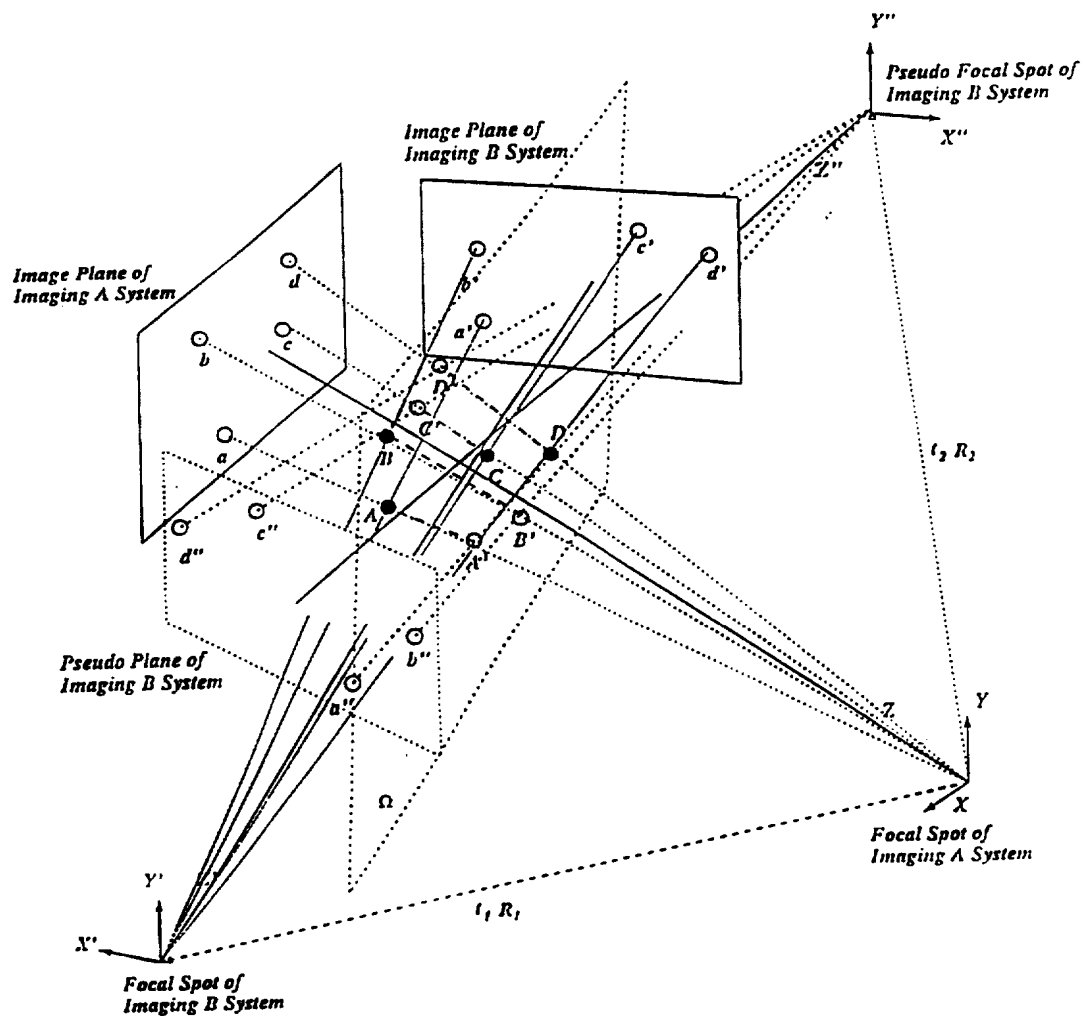


FIG. 7

INTERNATIONAL SEARCH REPORT

International Application No

PCT/US 97/10194

A. CLASSIFICATION OF SUBJECT MATTER
IPC 6 G06T11/00

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
IPC 6 G06T

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practical, search terms used)

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category *	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
Y	WAHLE A ET AL: "ASSESSMENT OF DIFFUSE CORONARY ARTERY DISEASE BY QUANTITATIVE ANALYSIS OF CORONARY MORPHOLOGY BASED UPON 3-D RECONSTRUCTION FROM BIPLANE ANGIOGRAMS" IEEE TRANSACTIONS ON MEDICAL IMAGING, vol. 14, no. 2, pages 230-241, XP000520934 see page 230, right-hand column, line 32 - page 231, right-hand column, line 14 see page 233, right-hand column, line 23 - page 234, left-hand column, line 16; figures 5-10	1-9,13, 15-18
Y	US 4 875 165 A (FENCIL LAURA E ET AL) 17 October 1989 see column 5, line 31 - line 45; claim 1 --- -/--	1-9,13, 15-18

☒ Further documents are listed in the continuation of box C.

☒ Patent family members are listed in annex.

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Y document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art.

& document member of the same patent family

Date of the actual completion of the international search

10 October 1997

Date of mailing of the international search report

22. 10. 97

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INTERNATIONAL SEARCH REPORT

International Application No.

PCT/US 97/10194

2. (Continuation) DOCUMENTS CONSIDERED TO BE RELEVANT

Category *	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
A	<p>WAHLE A ET AL: "A NEW 3-D ATTRIBUTED DATA MODEL FOR ARCHIVING AND INTERCHANGING OF CORONARY VESSEL SYSTEMS" PROCEEDINGS OF THE COMPUTERS IN CARDIOLOGY CONFERENCE, LONDON, SEPT. 5 - 8, 1993, no. -, INSTITUTE OF ELECTRICAL AND ELECTRONICS ENGINEERS, pages 603-606, XP000478369 see figures 1-4</p> <p style="text-align: center;">-----</p>	1-20

INTERNATIONAL SEARCH REPORT

Information on patent family members

Internal Application No

PCT/US 97/10194

Patent document cited in search report	Publication date	Patent family member(s)	Publication date
US 4875165 A	17-10-89	NONE	
